A study on high-precise magnetoencephalographic field-source estimation using the head model

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1 Introduction

Recently, the technique of the brain function imaging using Superconducting Quantum Interference Device (SQUID) technology has been developed with technical advancements of the SQUID [1, 2, 3, 4]. Imaging procedures consist of measurements of the external magnetic fields produced by electric phenomena due to neural activities in the brain, and the estimation of the magnetic sources from the measured magnetic fields. At present, the magnetic source is assumed as a current dipole, and the localization of point-like neural activity, appropriate for the current dipole assumption, is commonly performed. But the estimation becomes proportionally more difficult as the number of dipoles increases. Therefore, in the case that the sources are distributed extensively, as in epilepsy, their localization, by the conventional method mentioned above, is difficult to estimate. A precise estimation method for such sources is then necessary.

In this paper we present an estimation for location of the neuromagnetic source in the cortical region using MR Images [6]. MR images are used for limiting estimation region to cortex, making realistic head model, and analyzing acquired results. Magnetic fields of the brain are due to neural activites, and as there are many neurons on the cortex, we regard it as a prime candidate for source location. The cortex is extracted from MR Images. We assume that the points on the extracted cortex are current dipoles and all the acquired points are considered as candidate sources. We then perform the estimation by means of the MUltiple SIgnal Classification (MUSIC) method [3, 4, 5, 6]. The acquired results may be interpreted as candidate sources. We use realistic model instead of sphere model, and consider the effect of the conductivity and the shape of the head model in MEG. We apply this algorithm to real human brain somatically evoked magnetic field data.

2. Extraction of the cortex and conducting of realistic head model

The brain consists of the cortex and the medulla. The cortex is called the gray matter as it looks gray because of its many neurons. Neural activities are therefore concentrated in this layer, and we may regard it as an area where magnetic sources exist. So we extract this gray matter from MR images and define source locations to be within the extracted area.

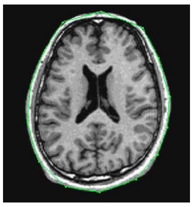
We remove the skull and the scalp from the MR images by fuzzy clustering to extract only the brain image [5, 6]. The outline is derived from this brain image to acquire the gray matter region. This derivation is performed by binarization using thresholding. However due to regions such as fissures or sulci, the brain has a complicated structure. Thus, if we use the brain image directly for the derivation of the outline, the gray matter boundary in these complicated regions cannot be acquired. To remedy this we adjust the threshold value and slightly erode the brain image. The outline is then acquired from this contracted image. Thus each point on the outline is regarded as representing a location between the brain surface and the medulla. This derivation is applied to all the slice images which include the brain. Acquired points are candidate dipole locations to be used in the estimation.

The estimation method requires two models. One is a head model, the other a magnetic sources model. Here we describe the head model. The model of head shape is important in the formulation of the relationship between the magnetic field and the magnetic source. At present the sphere model is widely used because of its simplicity. Using the sphere model, the formulation becomes relatively simple. But, it is well known that the real head shape is not a spherical shape. So a more realistic model of the head shape is preferable [2, 4]. We make realistic head model from MR images.

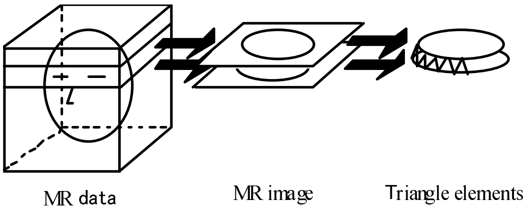
Here we deal with three layers: the scalp, the skull and the brain. We produce triangle elements between two slice images as shown in Fig.1. This procedure is applied to all images which include the scalp, the skull, and the brain, and we acquire a realistic model. Fig.1 shows the acquired models.

3. Estimation algorithm

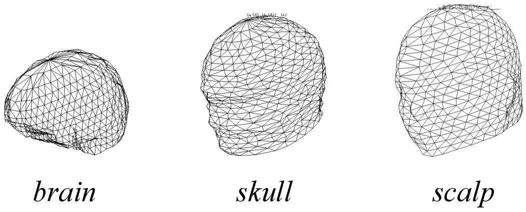
In the previous section we described the head model. Here we introduce a model of the magnetic sources and the formulation. Electric phenomena due to neural activities in the brain produces a magnetic field outside the head [1]. We assume that this generated field is due to current dipoles, and that the head is modeled as a realistic model. Moreover, it is assumed that the head model is divided into m different regions. Here we deal with three layers so m =3.



(a) MR images



(b) MR images and triangle elements



(c) Realistic model of the head

Figure 1 : *Head model*

Then the relationship between the magnetic field and current dipole can be described [2, 4] by

$$\mathbf{b}(\mathbf{r}) = \mathbf{b}_0(\mathbf{r}) - \frac{\mu_0}{4\pi}\sum_{j=1}^{m}(\sigma_j^- - \sigma_j^+)\cdot\int_{s_j} V(\mathbf{r}')\frac{(\mathbf{r}-\mathbf{r}')}{|\mathbf{r}-\mathbf{r}'|^3}\times d\mathbf{S}_j \tag{1}$$

where $\mathbf{b}(\mathbf{r})$ is a magnetic field at $\mathbf{r}$, and the "+"("-") superscript indicates the conductivity outside (inside) the j th surface. $V(\mathbf{r})$ is the potential at $\mathbf{r}'$ on the surface $\mathbf{S}_j$. $\mathbf{b}_0(\mathbf{r})$ represents the magnetic field produced by the primary current source in an infinite homogeneous medium. The second term represents the contribution of the volume currents which is seen to be equal to the magnetic field of a surface current distribution on $\mathbf{S}_j$.

Now we assume that there are M sensors, N dipoles and K time samples. Then the previous equation is represented using matrix notation, $\mathbf{B} = \mathbf{GQ}$. This notation indicates the linear relationship between the magnetic field and the dipole moment, and that dipoles have a fixed location for all measurement time [4]. $\mathbf{B}$ is an $M \times K$ matrix. $\mathbf{G}$ is an $M \times 3N$ matrix which stands for the relationship between the sensors, the dipole locations, and head model. $\mathbf{Q}$ is a $3N \times K$ matrix of dipole moments.

The estimation is generally performed by the parameter fitting in the least squares sense. But it becomes difficult when many dipoles exist. So we apply the MUSIC method [3, 4] to the spatio-temporal data $\mathbf{B}$. At first, the eigendecomposition of $\mathbf{R_B}$, which is the covariance matrix of the measured magnetic fields $\mathbf{B}$, is performed. $\mathbf{R_B}$ is decomposed as

$$\mathbf{R_B} = \frac{1}{K}\mathbf{B}\mathbf{B}^{\mathrm{T}} = [\mathbf{U}_s\mathbf{U}_n]\begin{bmatrix}\Sigma_s & \\ & \Sigma_n\end{bmatrix}[\mathbf{U}_s\mathbf{U}_n]^{\mathrm{T}} \tag{2}$$

where $\mathbf{U}_s$ and Σ_s are the matrix containing eigenvectors and the diagonal matrix whose diagonal elements are eigenvalues of the signal subspace, and $\mathbf{U}_n$ and Σ_n are those of the noise subspace, respectively. If the noise is the white Gaussian noise, the eigenvalues of the noise subspace are equal to the variance of the noise σ^2.

Next, we divide $\mathbf{R_B}$ into the signal and the noise subspaces using the acquired eigenvalues. In order to do so, we use the Minimum Description Length (MDL) criterion [5]. This criterion selects the separation point between the signal and the noise subspaces from the acquired eigenvalues automatically.

The estimation is performed by calculating

$$J = \lambda_{\min}\left(\mathbf{U}_{\mathbf{G}_i}{}^{\mathrm{T}}\mathbf{U}_n\mathbf{U}_n{}^{\mathrm{T}}\mathbf{U}_{\mathbf{G}_i}\right) \tag{3}$$

in the search region. $J = \lambda_{\min}(\cdot)$ is the minimum eigenvalue of the bracketed term, $\mathbf{U}_{\mathbf{G}_i}$ is obtained from the singular value decomposition (SVD) of matrix $\mathbf{G}_i$ which is an $M \times 3$ matrix for one dipole, and $\mathbf{U}_n$ is the matrix containing eigenvectors of the noise subspace. MUSIC requires the one dipole matrix $\mathbf{G}_i$, regardless of the true number of dipoles, i.e.. We have only to calculate (3) on each point in the search region. We utilize the cortex obtained from MR images for this search region. Each point on the cortex is used for calculation of $\mathbf{G}_i$. In previous work, on the cortex, we found that the area where (3) becomes small.

4. Experimental results

Among many suggested methods until now, the head model most widly used is a sphere model. However the shape of human head is different from sphere and the conductivity of the head is not uniform. As a basic study for accurate estimation, we present the effect of the conductivity and the shape of the head model on the estimation algorithm presented in section 3.

4.1 Effect of the conductivity of the head model

We performed computer simulations to present how the variation of the conductivity inside the head model affect the performance of the estimation algorithm. In the simulations, the number of triangle elements of brain, skull, and scalp were 1122, 1123, and 1118, respectively. The permeability was $4\pi \times 10^{-7}$ (H/m). The number of sensors was 66. We calculated the magnetic fields by locating a current dipole in (0.0, 0.0, 8.5) (cm). The moment was 1.0 (nAm) in the direction of X-axis.

We varied the conductivity values inside the head model under the following conditions.

- Condition (1)

The conductivity of the skull model is 0.003 (S/m) and the scalp model is 0.3 (S/m). Conductivity σ_b of the brain model is varied (where $\sigma_b = \sigma_0 C$, $\sigma_0 = 0.3$, C=0.25, 0.33, 0.5, 1.0, 2.0, 3.0, 4.0).

- Condition (2)

The conductivity of the brain model is 0.3 (S/m) and the scalp model is 0.3 (S/m). Conductivity σ_{sk} of the skull model is varied. (where $\sigma_{sk} = \sigma_1 C$, $\sigma_1 = 0.003$, C=0.25, 0.33, 0.5, 1.0, 2.0, 3.0, 4.0)

- Condition (3)

The conductivity of the brain model is 0.3 (S/m) and the skull model is 0.003 (S/m). Conductivity σ_{sc} of the scalp model is varied. (where $\sigma_{sc} = \sigma_2 C$, $\sigma_2 = 0.3$, C=0.25, 0.33, 0.5, 1.0, 2.0, 3.0, 4.0)

The experimental result of condition (1) is shown in Fig 2. From experimental result, we found that the conductivity of the head model have little effect on the estimation. The cost function shows a single distinct peak at the true position Z = 8.5 (cm). The estimation results with respect to the position and the moment of the current dipole were agree well with the true values. In the simulations under the conditions (2) and (3), we obtained the simular results to those of condition (1).

4.2 Effect of the shape of the head model

We performed computer simulations to present how the variation of the shape of the head model affect the performance of the estimation algorithm. In the simulations, the conductivity of brain, skull, and scalp were 1122, 1123, and 1118, respectively. The permeability was $4\pi \times 10^{-7}$ (H/m). The number of sensors was 66. We calculate the magnetic fields by locating a current dipole in (0.0, 0.0, 8.5) (cm). The moment is 1.0 (nAm) in the direction of X-axis. Number of triangle elements by each model is shown at table 1. We varied the number of triangle elements of the head model under the following conditions.

- Condition (1)

The skull model has 1123 triangle elements and the scalp model has 1118 triangle elements. The number of triangle elements of the brain was varied from #1 to #5.

- Condition (2)

The brain model has 1122 triangle elements and the scalp model has 1118 triangle elements. The number of triangle elements of the skull was varied from #1 to #5.

- Condition (3)

The brain model has 1122 triangle elements and the skull model has 1123 triangle elements. The number of triangle elements of the scalp was varied from #1 to #5.

The experimental result is shown in Fig 3. In comparison with the spherical model, we see that the realistic head model can improve the performance of the estimation algorithm. The true value of the dipole position Z = 8.5 (cm) was agree well with the estimation value, when the number of triangle elements varied.

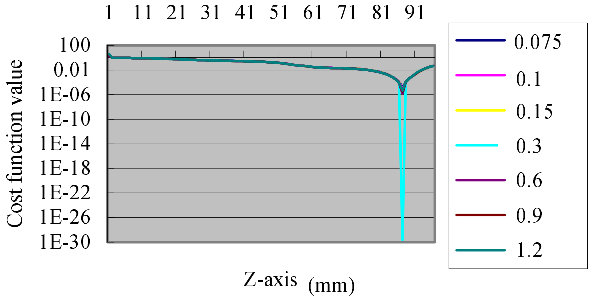
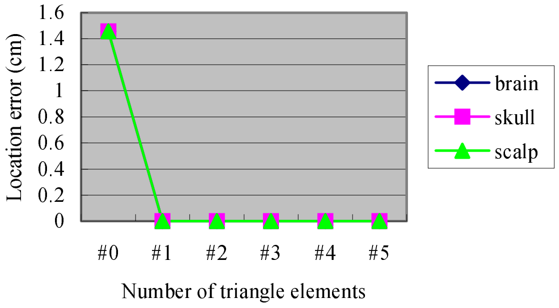


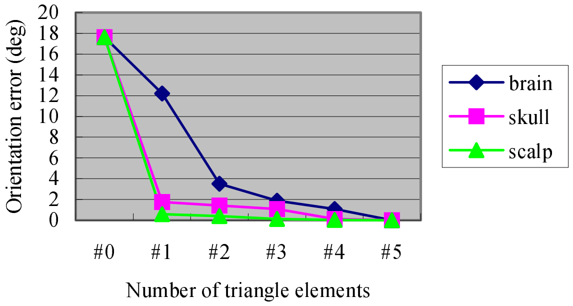
Figure 2: *The cost function values as the conductivity σ_b of the brain changed (σ_b = 0.075, 0.1, 0.15, 0.3, 0.6, 0.9, 1.2, σ_{sk} = 0.003, σ_{sc} = 0.3). The direction of the current dipole is an X-axis.*

Table 1: *Number of triangle elements by each model*

	Number of triangle elements
Brain	640(#1),752(#2),824(#3),1074(#4),1122(#5)
Skull	650(#1),786(#2),874(#3),1084(#4),1123(#5)
Scalp	628(#1),718(#2),862(#3),1002(#4),1118(#5)

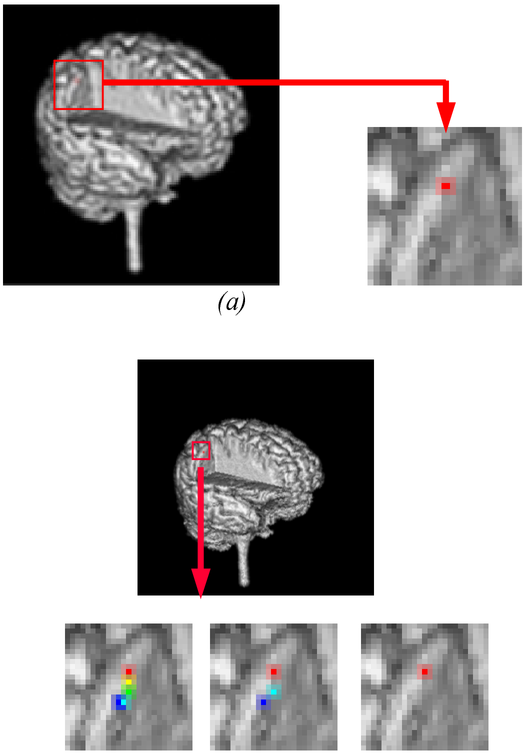


(a) Location error



(b) Moment error

Figure 3: *Plots of error values for with respect to the number of triangle elements.*



(a)

Brain *Skull* *Scalp*

(b)

Figure 4: *The integration image of MR image and estimation result.*
(a) The estimation results varying the conductivity values, (brain=0.075, 0.1, 0.15, 0.3, 0.6, 0.9, 1.2, skull=0.003, scalp=0.3 [S/m]).
(b) The estimation results varying the number of triangle elements, (Blue is #1. Light blue is #2. Green is #3. Yellow is #4. Red is #5.).

While the estimated value of the dipole moment approached the true value as the number of triangle elements of the head model was increased. From simulations results, we found that the performance of the estimation algorithm is more depended on the variation of the shape of the head model than that of the conductivity inside of the head model.

4.3 Application to SEF data

We applied the estimation algorithm to somatically evoked filed (SEF) data of a normal person. Electrical stimulation was given in the left wrist. The magnetic fields were measured with 66 sensors. The number of time samples was 150 on each sensor. It was applied to this SEF data analysis as well as section 4.1 and 4.2. Number of triangle elements by each model is shown at table 1.

Fig 4 shows the result synthesized with MR images. The variation of the location error with respect to the number of triangle elements is shown in Fig.5. The number of triangle elements of #6 and #7 are 1406 and 1632, respectively. From the experimental result, we found that the conductivity of the head model have little effect on the estimation, and that the effect of the brain model was bigger than those of scalp and skull models.

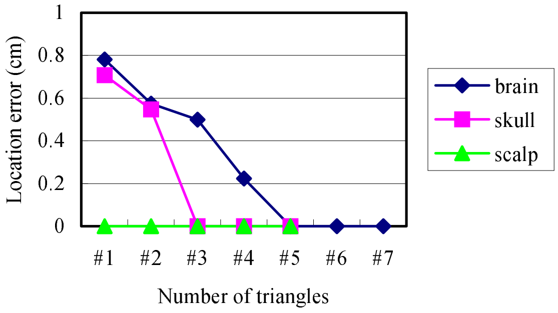


Figure 5: *Plots of error values for with respect to the number of triangle elements.*

5. Conclusion

We have presented an estimation algorithm for the location of magnetic sources using a realistic model. We constructed this head model from MR images of a human head. The head model consisted of three regions such as brain, skull, and scalp. We have performed the computer simulations and the somatosensory experiments to present how the head model affect the performance of the estimation method. From these experimental results, we found that the shape of the head model affected the performance of the estimation more than the conductivity inside the head model.

References

[1] J. Sarvas, "Basic Mathematical and electromagnetic concepts of the biomagnetic inverse problem," Phys. Med. Biol., 32, pp.11-22, 1987.

[2] M. S. Hämäläinen and J. Sarvas, "Realistic Conductivity Geometry Model of the Human Head for Interpretation of Neuromagnetic Data," IEEE Trans. Biomed. Eng., 36, pp.165-171, 1989.

[3] J. C. Mosher, P. S. Lewis, R. M. Leahyz, "Multiple Dipole Modeling and Localization from Spatio-Temporal MEG Data," IEEE Trans. Biomed. Eng., vol.39, pp.541-557, 1992.

[4] J. C. Mosher, R. M. Leahy, P. S. Lewis, "Matrix Kernels for MEG and EEG Source Localization and Imaging," Proc. ICASSP-95, Detroit, MI, 1995.

[5] K. Mino, N. Niki, I. Tamura, "Neuromagnetic Source Analysis with a 64-Channel SQUID System and MR Imaging," in Proc. IEEE Biomed. Image Analysis, Seattle, pp.255-260, 1994.

[6] K. Mino, N. Niki, N. Nakasato, T. Yoshimoto, "An Estimation Algorithm of Neuromagnetic Source Distribution Using MRI Information," IEEE Trans. on Nuclear Science, 44, pp.230-234, 1997.